Fast Spin-Echo

Joseph C. McGowan, PhD
CAPT, USNR

Associate Professor of Electrical Engineering

United States Naval Academy

Introduction

Imaging with T2-weighting is prescribed in nearly every diagnostic imaging protocol, owing to the fact that in many pathologic states, diseased tissue may be distinguished from normal tissue through apparent differences in T2. Conventional T2-weighted imaging carries with it a heavy time cost, however, as a relatively long time must be allowed between acquisitions in order to allow T1 recovery (and thus to avoid T1-weighting). Fast spin-echo imaging was designed to take advantage of the "dead time"between successive excitations, allowing true T2 contrast to be obtainedina fraction of the time required for a conventional scan.

Introduction to k-Space

K-space is a somewhat counterintuitive name but can be thought of as an array of numbers where MRI signals from the coil arestored. When k-space is "filled", the aggregate data can be processed viathe Fourier transform resulting in the image. The potentially confusing partis that there does not exist a one-to-one correspondence between a pointin k-space andan associated point in the image. Rather, each single pointin k-space contains information that is used to reconstruct the entire image.

K-space is reasonably simple to understand if oneadopts a black box approach for the Fourier transform, viewing it as thenothingmore than the mechanism used to convert the k-space map to the image.Addto this that the individual signals from the coil must find their placeinthe k-space grid of numbers. Now we see that a position in k-space isrelatedto somehow to time (when in the acquisition the voltage was measured)whilethe positions in image space are obviously related to space (position).Inboth cases (k-space and image) we are simply dealing with an array ofintensityvalues.

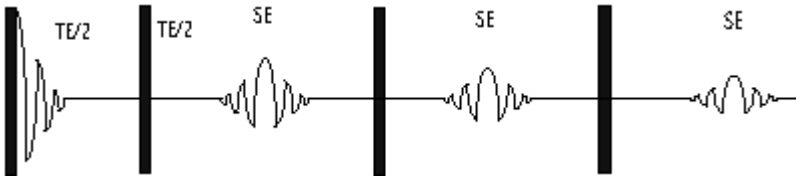
In order to obtain a line of k-space one completes one acquisition- a phase encode step- with frequency encoding during theacquisition. Usually, we are trying to capture a spin echo. To acquire thisdata we use the rf coil and some mechanism to measure the voltage at thecoil at manytimes during the acquisition window. These voltages are "writtendown" asa line of k space. So, one line of k-space is just a graphical representationof a single spin echo. When k space is full there exists enough informationto reconstruct the image, done of course with the (black box) Fourier transform.

Looking inside the black box, we see that the Fourier transform is a way of representing (breaking apart!) any function or lineshape (such as a spin echo) as a series of sine curves. The spin echo that we receive is a complicated function, as it contains signal at many different frequencies. We have arranged it this way by using gradients to do frequency encoding. These gradients established a correspondence between frequency and spatial position, so we know that if we can identify the contribution to the aggregate signal of a certain frequency, we can associate that with a position in space. This is exactly what the Fourier Transform does in the image reconstruction.

Gradients define position in k-space. The k-coordinate (or, the address of a point in k-space) is defined as : $k=(2^{)-1}\gamma Gt$ This kcoordinate is proportional to the area under the gradient time curve. Thereadout gradient controls kx, and the phase encode gradient controls ky.Thus, when the readout gradient is turned on, the kx-coordinate begins toincrease, and the "position" in k-space moves to the right, sweeping acrossa line of k-space. During the following acquisition, with the phase encodegradient incremented, the sweeping across will be repeated for all x valuesat a different y value.

Principles of Fast Spin-Echo imaging

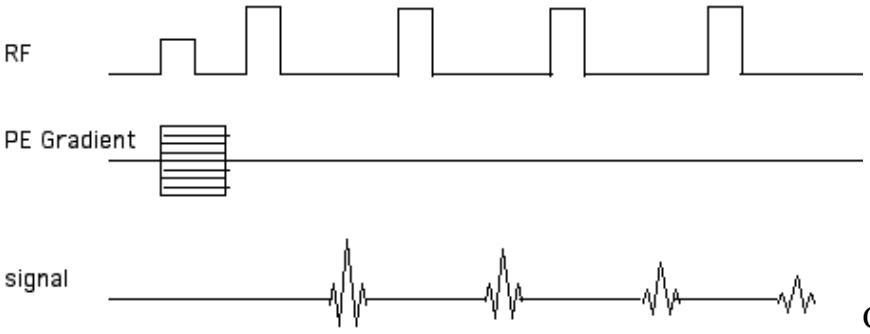
In the conventional spin-echo pulse sequence, thetransverse magnetization is refocused by the inversion (180û) pulse,so that an echo results at time TE. Following the spin-echo, the magnetizationproceeds to dephase (with time constant T2*) just as it did following theinitial excitation, with irreversible dephasing (pure T2 effect) continuingunaffected by theinversion and echo. It is straightforward to modify thispulse sequence withthe addition of a second inversion pulse following theacquisition of theecho, which will have the effect of once again reversingthe reversible spin dephasing, and creating another spin-echo. This secondecho will occur after an interval equal to twice the time delay between thefirst echo and the second inversion pulse. The process may be repeated indefinitelyas long as the residual signal (from pure T2 dephasing) is sufficient toallow detection. In practice,up to four evenly spaced echoes may be obtainedin this manner on standardclinical scanners. The pulses and echoes are depicted below for a seriesof three spin-echoes.



Spin-echo sequence for acquisition of three spin-echoes per excitation

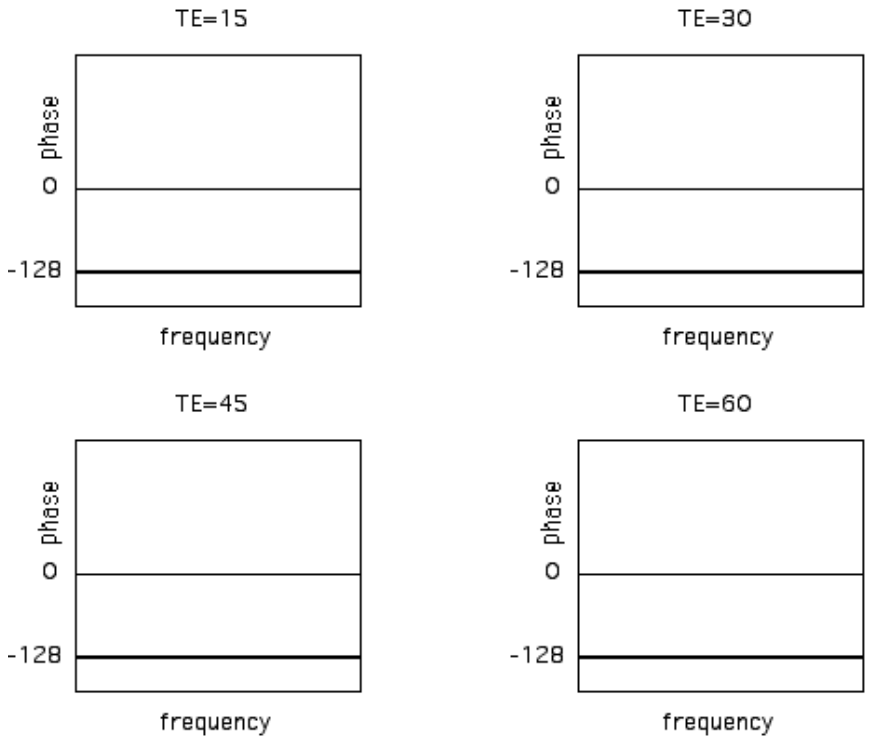
In conventional spin-echo imaging, the phase-encode gradient is applied only once, even if multiple echoes are acquired. Thus, using a sequence such as that shown above, three images will be acquiredfor each phase-encode step (each TR). The three images will demonstratecontrast according to three values of TE, corresponding to the time intervalsbetween the initial excitation and the three echoes. The magnitudes of theechoes will decrease with T2 , so that the later echo images will be on-the-whole darker than the early echo images, but each will posses unique contrast according to the TE of the image.

Of the multiple echo spin-echo technique we couldstate that several lines of k-space are acquired per TR, even though onlyone line of k-space per image is acquired. Fast spin-echo (FSE) is the logicalextension of technique of acquiring multiple lines of k-space during a singleTR. The difference in FSE is that instead of being used to construct severalimages, all of the lines of k-space obtained in the TR period are used fora single image.



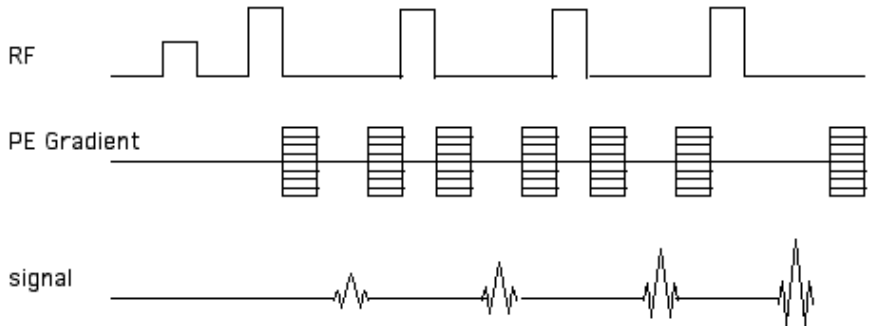
Consider a conventional spin-echo experiment with TR 2000 ms, TE of 15 ms with four evenly-spaced echoes acquired, and a 256 X 256 matrix and 1 NEX. This sequence would require 8:32 (2 s. X 256 PE steps) and would produce four images, with TE□s of 15, 30, 45, and 60 ms. Again, during each TR period one phase-encoding gradient would be applied and one line of k space would be acquired for each image. Also, each signal used for, say, the 15 ms TE image, would be acquired at 15 ms after the excitation. The sequence is diagrammed as follows and the k-space diagram for the first TR of this sequence is shown below.

Pulse sequence diagram for multiple echo spin-echo. The number of TR periodsrequired to acquire one or more images is equal to the number ofphase-encodevalues desired.

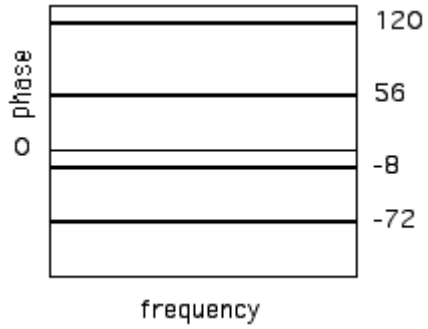


k-space diagrams for a multiple echo spin-echo sequence. During the first TR the line corresponding to phase-encode gradient of -128 is acquired for each of four images, with TE's evenly spaced multiples of 15 ms. Each k-space diagram corresponds to an image.

Now consider a sequence where each echo is acquired with a different phase-encoding gradient, that is, where four different values of phase-encoding gradient are employed for each TR. It is now possibleto construct a single image four times as fast, because 256 phase-encodevalues (or lines in k-space) take only 256/4 or 64 TR periods. Scan timehas been reduced to only 2:08 for the single image in this FSE acquisition. Here are the sequence and k-space diagrams for this method.



Pulse sequence diagram (partial) for one TR of fast spin-echo, with echo train length equal to 4.



k-space diagram for the first TR of the above pulse sequence. Valuesof phase encode gradient are (in order) 120, -72, 56, -8. This orderingaccounts for the increase of signal intensity apparent on the pulse sequencediagram. Ordering such that the low phase-encode (high signal) echoes areacquiredlate in TR contributes heavy T2 -weighting.

At this point one may observe that all values of phase encode gradient can be acquired in this way but the signals from four different TE periods are simply being mixed together. What is the TE of the resultant image? The question is answered simply if one recalls that different regions of k-space contribute unequally to the final image. Remember that the area of k-space where the phase-encode gradient values are small or zero contribute the highest signals and are most important to the contrast of the image.Conversely, the high values of phase-encode gradient (high spatial frequencies)give edge definition but are less important to contrast. Thus, to acquirean image with contrast like that of, say, an 80 ms TE, one must arrange forthe low spatial frequency lines of k-space to be acquired 80 ms followingexcitation. The rest of k-space is filled in using the other echoes, whichthen have less influence on the contrast. For this reason the TE of a FSEsequence is given as effective TE.

Operator-controllable parameters in FSE imaging:

Besides the standard variables of the conventional spin-echo sequence, FSE requires the selection of echo train length (ETL), or the number of echoes acquired per excitation. Commercial scanners mayoffer ETL□s from 2 to 16. Selection of a value for this variable determinesthe time savings of the FSE sequence over a comparable spin-echo sequence,as was seen in the above example (ETL=4, time savings =factor of four). Theoperator may also elect to acquire dual echoes with the FSE technique. Thisoption, as expected, doubles the acquisition time, all else being equal. The time between echoes is

termed echo spacing and is also directly or indirectly under operator control. A consideration with FSE may be the number of slicesobtainable. Since FSE makes more efficient use of the entire TR time by fillingit with additional echoes and data acquisition, there may be relativelylesstime for encoding additional slices. In this case concatenated acquisitionsmay be used to cover the necessary anatomy. The following relationship isvalid:

$$T_a = \frac{N_{pe} \bullet NEX \bullet TR}{ETL}$$

where Ta is acquisition time, Npe is number of phase encodes, and ETL is echo train length.

Methods for acquiring dual-echo studies with FSE

Two-echo FSE sequences may be prescribed to obtain both proton-density and T2 weighted contrast in a manner analogous to conventional dual echo spin-echo. The images may be acquired in any of several ways.Of course, the normal FSE sequence may simply be run at each desired valueof effective TE to provide full flexibility of scan parameters and echo times. However, it may be desirable to obtain images representing both TE□s in the same scan. This may be done by concatenating scans in what is termed by GE as "fast variable echo". Another way to perform a dual echo acquisition is by splitting the echo train and assigning the early echoes to the one(PD-weighted) image and the late echoes to the other (T2-weighted image). Clearly, this arrangement provides for equal imaging time, all other parametersequal. There is some restriction on the setting of TE, as the first valueof TE mustoccur during the first half of TR, while the second TE must bein the secondhalf of TR. A possible advantage of this arrangement is thatthe effectiveTE images will more faithfully replicate conventional imagesat equivalentTE, as the actual echoes from which the signal is acquiredwill be closerto the desired TE value. Split echo train acquisition mayalso be effectivein reducing blurring artifact.

Advantages of FSE over conventional spin-echo

The advantage of FSE over conventional spin-echo is, clearly, the speed with which T2 weighted contrast may be acquired. Thereductions in imaging time are significant in that they make certain examinationsfeasible (i.e., within the normal capacity of a patient to remain still)which would not be otherwise. For example, ultra-long TR values can beprescribed to develop appropriate T2 weighting in tissues with longer T2 values (suchas in pediatric brain). Ultra long TR□s may also beused to enhancemyelographic effects, improve SNR, yield proton density informationin tissueswith long T1, and increase the number of slices available. LongerTE valuesare also available for additional T2 -weighting over what is possiblewithconventional spin-echo imaging.

The recent literature contains many examples of studies comparing FSE with conventional spin-echo imaging, and in many cases FSEprovides equivalent clinical value with a much smaller time expenditure. For example, one group found that long echo train length T2 weighted 3D-FSEsequence enabled the detailed visualization of the tiny structures of theinner ear and the internal auditory canal within a clinically acceptablescan time (European Radiology. 6(3):369-74, 1996). Another group combinedFSE with gradient echo techniques and reported accurate detection and localizationof labral injuries(Radiology. 200(2):519-24, 1996 Aug). Still others substitutedFSE for conventional spin-echo imaging, giving up proton-density weightedimages, and reported "no diagnostically relevant" difference between FSEand conventional spin-echo imaging of pediatric brain (Pediatric Radiology.26(4):259-64, 1996). In a study of forty-eight adult patients with low-backpain, it was concluded that "a rapid, two-plane, single-echo fast spin-echosequence protocol is adequate to detect potentially significant degenerativedisease of the lumbar spine. Its 2.5-min acquisition time allows a completepatient study (including patient preparation) to be performed in less than10-15 min" (American Journal of Roentgenology. 166(4):909-16, 1996 Apr).In another study comparing lesion conspicuity on head and neckimaging, FSEwas found to give equal or superior contrast to conventionalSE (AmericanJournal of Neuroradiology. 15(4):767-73, 1994.)

Differences between FSE and conventional spin-echo

It is now generally held that contrast in FSE imaging may be viewed as equivalent to that in conventional spin-echo imaging for many diagnostic purposes. This acceptance is reflected in the large numberof clinical protocols at the Hospital of the University of Pennsylvania whichemploy FSE. There are, however, important differences between FSE and conventionalspin echo imaging, which should be recognized and considered. Fat is generallybright on FSE images as compared with conventional spin-echo, an observationwhich has been attributed to the absence of the effects of J-modulation.(HenkelmanRM et al, Why is fat bright in RARE and FSE images? J Magn ResonImaging 2:533-540,1992) Magnetization transfer effects due to the excitation of adjacent slices maybe significant in FSE, due to the large number ofinversion pulses, whichappear to an adjacent slice as pulsed off-resonanceexcitation.(Dixon WTetal, Incidental magnetization transfer contrast affectsordinary multisliceimaging. Magn Res Imaging 8:417-422, 1992) This effect will tend to enhancecertain types of contrast, such as the contrast betweengray and white matter.Additionally, it has been suggested that FSE (particularlyat high valuesof ETL) may exhibit decreased sensitivity to diffusion-mediatedsignal loss,and thus may not be the method of choice for detection of acutehemorrhage. The opposite view has been advanced by others (Jones K. et al,Brain hemorrhage: evaluation with fast spin-echo and conventional dual-echospin-echo images. Radiol 182:53-58, 1992), who hold that there is no differencein hemorrhagic lesion conspicuity.

Problems with FSE imaging

In general, the advantages of FSE imaging are sufficient to justify its general diagnostic use. In certain situations, however, disadvantages must be considered. One disadvantage is the relatively high SAR generated by this sequence, which may limit the number of allowable slices. This limitation is severe in whole-body imaging at higher fields such as 4 Tesla.

Some artifacts specific to FSE may arise as a consequence of the discontinuous traversal of k-space. These may include blurring, edge enhancement, and ringing. Blurring arises when the T2 decay is such that the later echoes do not contain enough signal to be detected over the inherent noise. If high spatial frequencies are assigned to these echoes, the contrast will not be severely affected, but the apparent resolution will. The degree of blurring is estimated by use of the point spread function, an expression which describes the extent to which signal from a point source will appear in adjacent pixels. The line width (full width at half max) of the point spread function in FSE is:

$$FW_{\frac{1}{2}} = \frac{\Delta y \bullet 2}{\pi} \frac{ETL \bullet E_s}{T_2}$$

with E_s representing echo spacing and ETL echo train length. The pixel size is given as Δy. As an example, for a substance with a TE of 60ms, and assuming an ETL of 16 and spacing of 20 ms, the line width of the point spread function is greater than 3, indicating that a point source of one pixel would be detected over 3 pixels. This admittedly extreme example demonstrates the need for consideration of where resolution and edge definition is important for diagnosis.

The opposite effect may present itself when the high spatial frequency echoes are relatively strong compared with the low, such as when acquiring a proton-density weighted image. In this case edge enhancement may result. One should be concerned with this effect when a long T2 tissue is next to a short T2 tissue (such as fat).

These artifacts are all reduced by reduction of ETL or echo spacing. Of course, lowering ETL raises imaging time and reducing echo spacing may carry a SNR penalty if the bandwidth for acquisition must be adjusted. In a study designed to examine the interaction of echo train length and interecho spacing and their effects on image quality and contrast in FSE imaging of the cervical spine, it was reported that *increasingecho train length increased homogeneity and high intensity of cerebrospinalfluid signal and reduced acquisition time; however, it decreased the signal-to-noise ratio of cerebrospinal fluid and cord and increased blurring, and, to a lesser extent, edge enhancement, and "truncation-type" artifact. Increasing interecho space permitted the use of longer echo times but minimally decreased contrast and signal-to-noise ratio of cord and cerebrospinal fluid. In addition, increasing echo spacing increased blurring, edge enhancement, truncation-type, magnetic susceptibility, and motion artifacts*(American

Journal of Neuroradiology. 14(5):1203-13, 1993). These results are consistent with the relationships given above.